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Load-Speed Interaction Effects on the Biomechanics of Backpack Load Carriage

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Summary

We biomechanically examined how backpack load and walking speed interact in their effects. 16 males walked under all 12 combinations of 6, 20, 33, and 47 kg backpack loads and 1.17, 1.33, and 1.50 m/s walking speeds. Generally, the effects of load were consistent over the speeds, and the effects of speed were consistent over the loads. Ground reaction forces and impulses, joint forces, muscle torques, muscle electrical activity and backpack acceleration increased when speed and/or load increased, likely increasing the probability of fatigue and injury. As load increased, percentage of stride in double-support and time of toe-off increased, and maximum hip angle decreased, likely improving stability and reducing stress on the musculoskeletal system. However, increases in walking speed tended to cancel these adaptations. At the lower speeds but not the highest one, stride frequency increased and stride time decreased when the load increased from 33 to 47 kg. Downward impulses for the major lower body joints increased with load carried, but decreased as walking speed increased. At the 1.33 m/s speed, but not at 1.50 m/s, a gait adaptation resulted in a less-than-expected impulse increase when the load increased from 33 kg to 47 kg. At the fastest walking speed, the volunteers could not further increase stride frequency to reduce stride length, increase stability, and reduce potential lower body stresses. Thus, it appears that soldiers should avoid, if possible, walking faster than 1.33 m/s (4.8 km/hr; 3.0 mi/hr) when carrying backpack loads approaching 47 kg (100 lb).

Introduction

While there has been much research on the biomechanics of human gait, only a small proportion of such research has specifically addressed load carriage. In 1981, Pierrynowski, Norman, and Winter (30) used cinematography to investigate variation in the mechanical energy levels of the body segments and efficiency of volunteers carrying five different backpack loads. Kinoshita and Bates (24) compared the effects on ground reaction forces of a standard backpack vs. a two-pack system, the latter of which distributed the load equally between the front and back of the volunteers. In another study, Kinoshita (23) reported significant changes from unloaded body posture and gait pattern when loads of 20% and 40% of body weight were carried, but less deviation from normal walking with a front/rear pack system than a standard backpack. Our laboratory compared the effects of a load carriage system that distributed the load between the front and back of the torso to the effects of a standard backpack on walking posture both before and after a fatiguing maximal speed 20 km road march (12, 16). We also compared various load carriage systems as to walking and running biomechanics among both male and female soldiers (17, 18). Electromyography has been used to evaluate muscle activity during walking, especially in the lower extremities (4, 6, 27). Yet most studies of load carriage have been physiological rather than biomechanical and have focused on metabolic response (2, 9, 11, 14, 21, 28, 32).

Many investigators have biomechanically analyzed unloaded human locomotion, using methodology that can be applied to the study of load carriage. They evaluated stride length (35, 36), joint forces and moments (5, 7, 22), joint ranges of motion (26), path of the center of pressure on the foot (15, 38), mechanical power (25, 41), external work (13), timing of gait events (38), braking impulse (29), and the effects of speed on mechanics (31). Electromyography (EMG) has been used to determine which muscles are involved in a physical activity, estimate their contraction intensity, and determine the muscle contraction sequence (3, 10, 33, 34, 40). Stulen and De Luca (37) used EMG frequency analysis to gain insight into the effects of fatigue on motor unit recruitment patterns.

Most of the commercial and military backpack systems and other load carriage equipment available today have not been tested biomechanically. Application of quantitative biomechanical evaluation to loaded human locomotion can potentially contribute to the effectiveness of equipment evaluation and design. Thus, we undertook the study upon which this report is based in order to gather information on the effects of backpack load and walking speed on gait kinematics and kinetics. The goal was to expand the knowledge upon which recommendations concerning pack systems, physical training programs, and load carriage technique are based. It was anticipated that this could ultimately benefit people who engage in load carriage for whatever purpose by increasing load capacity and transport speed, lessening the likelihood of injury, improving efficiency, and decreasing perceived level of difficulty. We published two technical reports based on the study, one addressing the effects of backpack weight (19) and the other the effects of walking speed (20) on gait biomechanics. The purpose of this report is to provide a closer look into how load and speed interact in their effects.

Methodology

Volunteers

Testing took place at the biomechanics laboratory of the U.S. Army Research Institute of Environmental Medicine, in Natick MA. There were 16 male volunteers for the experiment, including military volunteers assigned for a tour of duty to the U.S. Army Natick Soldier Center, soldiers recruited for temporary duty as test volunteers, and military and civilian employees of the U.S. Army Research Institute of Environmental Medicine.

A nomogram for repeated measures (8) was used to estimate the sample size. To find the number of volunteers needed, a line was drawn from the inter-trial correlation coefficient through the desired effect size to the sample size scale. Inter-trial correlation coefficients of most dependent variables analyzed in the biomechanical study of load carriage were available from pilot study. The inter-trial correlation coefficients for most of the variables examined were higher than 0.60. For an inter-trial correlation coefficient of 0.60 with a moderate effect size of 0.5 and a two-tailed alpha level of 0.05, 13 volunteers were needed. Sixteen volunteers were recruited in order to provide for data lost by equipment malfunction or to make up for volunteers who might terminate testing prematurely.

Instrumentation

Force Platform System. Information from the force platform included forces exerted by the feet in the vertical, front-back, and left-right directions relative to the walker as well as the location on the platform of the foot center of pressure. Knowledge of the latter was essential for calculation of the moment about the ankle joint due to ground reaction force, and the subsequent calculation of torques about the knee and hip.

The model LG6-1-1 force platform from Advanced Mechanical Technology Incorporated (Newton, MA), measuring 0.61 by 1.22 m, was mounted on a steel frame to keep it rigid and isolated from external vibrations that might cause spurious output signals. The system was designed to emit voltage signals proportional to forces and torques exerted on the plate's surface, which include forces in the vertical, front-back and left-right directions and torques around orthogonal axes through the center of the plate oriented in the latter three directions. Center of pressure was calculated from the forces and torques, as specified in the AMTI force platform manual (1). The force platform and walking surfaces were made flush by locating the force platform at the center of a custom-built 15 m long wooden walkway. A model SGA6-3 amplifier system, designed for computer data acquisition, contained a six-channel amplifier with switch-selectable gains of 1000, 2000, and 4000 for each channel. Each channel also had a selectable low-pass filter with a 10 Hz or 1,050 Hz cutoff frequency and selectable precision bridge excitation voltages of 2.5, 5, or 10.

Accelerometer. A model EGAXT3-84-c-100 tri-axial accelerometer (Entran Devices, Fairfield, NJ) was mounted in the pack during load carriage. It emitted voltage signals proportional to pack acceleration in three orthogonal directions. This temperature compensated strain gauge accelerometer measured accelerations in the range of ±100 g in the vertical, left-right, and front-back directions. Built-in over-ranging protection prevented

damage to the device. Because of a very high resonant frequency of 1,700 Hz, the accelerometer did not distort the accelerations characteristics of human movement.

Cinematography System. One LOCAM II camera from Redlake Corp. (Morgan Hill, CA), capable of filming speeds up to 500 Hz, was used to film the volunteers during load carriage. A frame rate of 60 Hz was used for this experiment to capture the body movements of interest. The camera incorporated a timing light that placed markers on the edge of the film every .01 sec to allow checking of film speed. A model 12-0101 battery pack permitted use of the camera away from AC power outlets. Model 9003-0001 floodlights (1000 watts) from Colortran (Burbank, CA) and model 18001 Mini-Mac photoflood lamps (1000 watts) from Bardwell & McAlister (Hollywood, CA) provided illumination.

For analysis, developed films were projected with an M-16C projection head from Vanguard Instrument Corp. (Melville, NY) onto an ACT23 digitizing table from Altek Corporation (Silver Spring, MD). The projector allowed one frame of the film to be seen at a time. Specific frames could be referenced using a digital frame counter. The digitizing table had a resolution of .01 mm and was connected via its controller to a model 486-33 IBM-PC compatible computer from Club American Technology Inc. (Fremont, CA). The experimenter used a pointing device to identify the major joint centers of the body on the film image. The digitizing device sent table coordinates of the joint locations to a computer, where programs processed the coordinate information to calculate kinematic variables that included body segment positions, velocities, and accelerations. The volunteer's body mass and data from a force platform were processed along with the kinematic data to produce kinetic information, which included the forces and torques at each body joint.

Electromyography System. "Utah" model surface electrodes with integral preamplifiers and band pass filtering systems from Motion Control Inc. (Salt Lake City, UT) were used to record muscle potentials from the shoulder, back and legs. Each electrode was factory calibrated, with individual gains ranging from 340 to 380. Although the gain was slightly affected by the frequency of the signal being amplified, the variation in gain for signals between 60 and 500 Hz was within 2% of the range. The bandwidth of the preamplifier was 8 Hz to 33 KHz. The high input impedance of the electrodes made it unnecessary to abrade the skin or use electro-conductive jelly.

Computerized Data Collection System. The data were sent to a model 486-33 IBM-PC compatible computer from Club American Technology Inc. (Fremont, CA), including six output signals from the force platform, three from the accelerometer, six from the muscle EMG electrodes, and one from the event marker, for a total of 15. The signals were fed into a model DAP1200/2 data acquisition and analog-to-digital converter board (DAP) from Microstar Laboratories Inc. (Redmond, WA) mounted in an expansion slot in the computer. The DAP combined analog data acquisition hardware with a 16-bit microprocessor and a real-time multitasking operating system. It had 16 channels, each of which could be specified in software as single-ended or differential.

The inputs to the DAP were voltages, which the board converted to numbers. The board could perform computations on the resulting numbers before the information was sent to the computer, making data processing very fast. The gain factor was independently software selectable for each channel, with possible values of 1, 10, 100, and 1,000. Allowable voltage input ranges with unity gain were 0 to 5 V, -2.5 to +2.5 V, -5 to +5 V, and -10 to +10 V. Maximum sampling rate was 50,000 per second. The sampling rate for this experiment was 1,000 Hz for all the channels except for the EMGs. Two logical channels operating at 1,000 Hz each were used for each EMG hardware channel, so that the actual sampling rate was 2,000 Hz per EMG channel.

Backpack. A backpack (Figure 1) was specially designed for the experiment, using a standard U.S. Army ALICE external pack frame as a base. Two metal shelves were added to the frame. On the bottom shelf was mounted a metal box containing the accelerometer, a terminal for the EMG electrodes, and a junction for a multi-conductor cable through which output data could be sent to the analog-to-digital converter board mounted in the computer. The top shelf of the pack was designed to hold weights so that the intended experimental loads could be carried in the pack. The weights were in the form of lead bricks and rectangular iron plates.
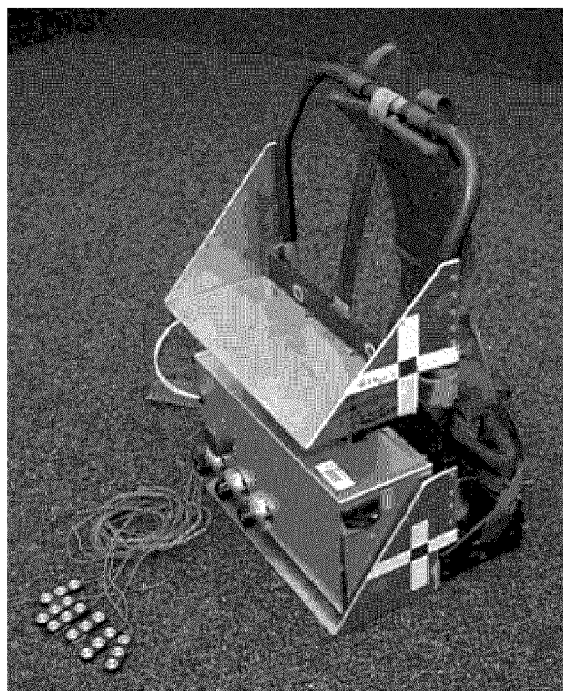
Figure 1. The instrumented backpack used in the experiment

An effort was made to match as closely as possible the location of the vertical center of mass of the experimental pack and an ordinary backpack. A pack loaded in standard fashion was balanced on a straight edge to locate its vertical center of mass. The weights were then arranged on the experimental pack in such a manner as to match the vertical center of mass location of the standard pack. Blocks of stiff foam were used as spacers on the shelf under the weights to make sure all of the pack loads had the same center of mass.

Two tape markers were placed on the side of the experimental pack so that the pack's position could be determined throughout a filmed trial by digitizing. The location of the actual pack center of mass relative to the markers was measured and recorded for use by the film analysis computer program.

Speed Cuing Device. A device to pace the volunteer's walking speed was designed at the U.S. Army Research Institute of Environmental Medicine and fabricated at the U.S. Army Soldier Systems Center in Natick, MA. It was based on a motor-driven cord marked with alternating light and dark bands that traveled around two pulley-wheels spaced 8 m apart. The speed of the cord was set using a dial. A digital display enabled cord speed to be set to the nearest 0.01 m/s. During an experimental trial, the device was oriented alongside the volunteer so that the visible part of the cord traveled in the direction the volunteer walked. The volunteer walked straight ahead while maintaining a peripheral view of the moving cord, which cued the appropriate walking speed.

Experimental Procedures

Independent Variables. Two independent variables were tested, backpack load and locomotion speed. The experiment was designed to test subjects under all 12 possible combinations of 4 backpack loads (6, 20, 33, and 47 kg) and 3 walking speeds (1.1, 1.3, and 1.5 m/s). The load of 6 kg was chosen because it was the weight of the backpack itself. The volunteers had to carry the pack even in the lightest load condition because the pack contained an EMG terminal as well as an accelerometer. The load of 47 kg was selected as a very heavy load that may be carried by serious backpackers and soldiers. The other two loads were equally spaced between the 6 and 47 kg loads. The 3 selected walking speeds can be respectively characterized as slow, medium, and fast.

Dependent Variables. The following variables were calculated from the vertical, front-back and left-right forces exerted by the feet on the force platform:

    a.   heel-strike and push-off peak forces (N)
    b.   time of occurrence of heel-strike and push-off peak force (percent of stride time)
    c.   peak and average front-back and mediolateral forces (N)
    d.   positive and negative vertical, front-back and mediolateral impulse per stride (N sec)

Film analysis allowed calculation of the following:

    a.   joint ranges of motion for the hip, knee, and ankle (radians)
    b.   joint torques for the hip, knee, and ankle (N·m)
    c.   joint forces at the hip, knee, and ankle (N)
    d.   stride length (m)
    e.   stride frequency (strides/min)
    f.   single-support time (percent of stride time)
    g.   double-support time (percent of stride time)
    h.   body segment and center of mass position, velocity and acceleration

EMG analysis allowed calculation of the following:
    a.   peak and average muscle activities for the trapezius, spinal erector, quadriceps, hamstrings, gastrocnemius, and tibialis anterior muscles (uV)
    b.   timing of activation for the muscles listed above

Accelerometer data analysis allowed calculation of the following:
    a.   peak accelerations of the backpack in the vertical, front-back, and left-right directions (g)
    b.   timing and directions of the accelerations

Test Trials. All volunteers were orally briefed on the purpose, risks, and benefits of the study, after which they signed informed consent documents. Electrodes were attached to the volunteers' skin with adhesive tape after the skin was cleaned but not abraded with rubbing alcohol and a gauze pad. Electrodes were placed over the following muscles using anatomical landmarks according to the recommendations for standardized electrode positions (42):

    - trapezius (elevates the shoulders, resists shoulder depression under the weight of the backpack)
    - lower erector spinae, L4/L5 level (extends the back, resists forward movement of the trunk due to backpack weight and inertia)
    - rectus femoris (extends the knee and flexes the hip during locomotion, helps lift the weight of body and backpack during the stride)
    - biceps femoris (flexes the knee, extends the hip)
    - tibialis anterior (works eccentrically to control the speed of foot plantarflexion so that the foot doesn't hit the ground too quickly)
    - gastrocnemius (plantarflexes the foot, helps lift the weight of body and backpack during the stride)

The volunteers performed their test trials (Figure 2) while wearing shorts and military boots. Prior to data collection, reflective tape markers were placed on the right side-view joint centers of the ball of the foot, ankle, knee, hip, shoulder, elbow, and wrist. Volunteers then donned the loaded backpack. Trials consisted of walks of no more than 15 m across the force platform in the camera field of view. Each volunteer was given practice trials to adjust walking speed and starting position so that the right foot landed squarely on the force platform as the volunteer walked across it. Occasionally, trials had to be repeated if the volunteer did not walk at the appropriate speed or did not place the foot completely within the confined of the force platform. A volunteer performed no more than nine trials in a test session (1 load x 3 speeds x 3 trials), with a maximum of two test sessions per volunteer per day (one in the morning and one in the afternoon). The volunteers were to walk at 1.1, 1.3, and 1.5 m/s corresponding to slow, medium, and fast walking with a backpack load, visually cued by the

specially designed speed-cueing device running alongside the volunteer. However, later cinematographic analysis revealed that their actual speeds were respectively 1.17, 1.33, and 1.50 m/s (4.2, 4.8, and 5.4 km/hr; 2.6, 3.0, and 3.4 mi/hr), which still can be characterized as slow, medium and fast backpack load carriage speeds. Subsequent to this experiment, an electric-eye speed-trap system was added to the experimental methodology to provide immediate feedback as to whether the volunteer walked at the cued speed. Each volunteer carried a different load on each test day resulting in a total of 36 acceptable trials over four test sessions. Occasionally, a trial had to be repeated if the volunteer's foot did not land directly on the force platform. Adequate rest periods were allowed between trials to avoid fatigue as a confounding factor. Each trial lasted no more than 15 seconds, so total exercise time per day was minimal.

Figure 2. The experimental setup. For the trials, the volunteers wore boots.

Data Processing. Data were collected and analyzed on the computer. Programs in the C++ computer language, specifically written for the study collected the digitizing table coordinates from each frame of film, as well as the data from the six force platform channels, the three accelerometer channels, and the six EMG electrodes, all converted from analog signals to numerical information by the A/D board. Other programs performed the processing necessary to compute records of dependent variable values over the stride. A large statistical file then was created which contained key variables describing the gait patterns of all the volunteers.

The EMG data underwent digital-to-RMS conversion (33) and other interpretive procedures. The vertical and horizontal forces determined from the force platform divided by the weight of body-plus-load gave vertical, mediolateral and front-back accelerations of the system center of mass. Mathematical integration of the accelerations yielded velocities.

Digitizing. Of the 3 trials of each volunteer per load-speed combination, data from the one closest to the target walking speed was selected for statistical analysis. An experimenter obtained the x-y image coordinates of each marker on a volunteer's body over a full stride by projecting the film one frame at a time on the rear side of the translucent digitizing table and sequentially placing the cross-hairs of a transparent mouse-like device over the center of each joint marker image. When the experimenter pressed a button on the device, the x-y digitizer table coordinates of the marker were sent to the computer. A custom-written Borland C++ computer program collected film data from the digitizing table via an IEEE-488 interface board (Capital Equipment Corp., Burlington, MA) installed in one of the computer's expansion slots. The program drew a stick figure of the

volunteer on the computer screen as the film was digitized to allow immediate detection and correction of gross digitizing errors. The computer displayed the name of each joint as it was to be digitized.

The ball of the foot, ankle, knee, hip, shoulder, elbow, wrist, and earlobe of the right side of the volunteer were digitized. The first frame digitized was 11 frames before the frame at which the right heel passed the back of the left lower leg. The last frame digitized was 12 frames after the right heel again passed the back of the left lower leg. This centered the gait data at right heelstrike, giving the best possible film images of the entire stride. The extra frames digitized at the beginning and end of the stride were needed for mathematical data smoothing and to ensure that a full stride was recorded. Before processing the film images from a given trial, the experimenter digitized the images of the four corners of the force platform, which were later used to calculate the film coordinates of the center of pressure, needed for the kinetic analysis.

Data Smoothing and Interpolation. The digitized film data were smoothed using Fourier analysis and Digital Filtering subroutines contained in Software for Science and Engineering Tools IPC-TC-006 (Quinn-Curtis, Needham, MA). The smoothed data were then processed with a cubic spline curve-fitting subroutine from the same software library to produce 101 interpolated frames for one full stride representing 0% to 100% of the time of a full stride. Thus, the results for each volunteer were in terms of percentage of stride. The actual time between interpolated frames was unique to each trial and was later used to calculate actual velocities and accelerations of the body segments and center of mass.

The mass, center of mass, and moment of inertia of each body segment were estimated using tables of standard body proportions based on dissection of cadavers (39). Because both heel-strike and toe-off were visible in the films and on the display of force platform data, these two points were used to time-synchronize film and force platform data. The EMG and accelerometer data were already time-synchronized with the force platform data because the computer's analog-to-digital converter board concurrently digitized them all. The foot's center of pressure location on the force platform's surface was calculated for each trial from force platform data using equations provided by the force platform's manufacturer (1). Joint moments and forces for the lower extremity were calculated using segment-by-segment kinetic analysis (39).

System of Postural Analysis. To analyze posture throughout the stride, the system of sagittal plane body angles shown in Figure 3 was used, in which:
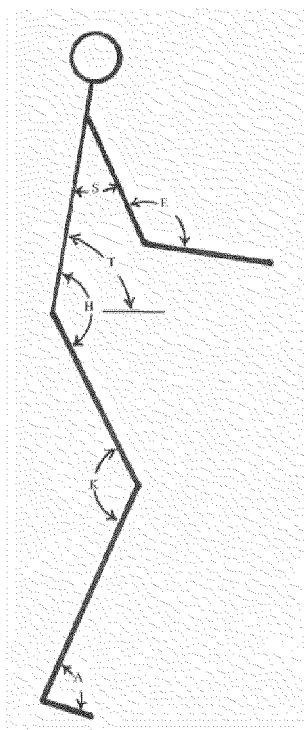
A = Ankle angle: the absolute ventral angle between foot and shank. Because the foot segment endpoints were the lateral malleolus and ball of the foot, when the bottom surface of the foot was at 90° relative to the shank, the ankle angle was about 120°.

K = Knee angle: the absolute dorsal angle between shank and thigh.

H = Hip angle: the absolute ventral angle between thigh and trunk.

T = Trunk angle: the ventral angle between the trunk and a horizontal line.

E = Elbow angle: the absolute ventral angle between upper arm and forearm.

S = Shoulder angle: the angle between upper arm and trunk (plus means upper arm is in front of the trunk; minus means upper arm is behind the trunk).

Figure 3. The system of sagittal plan body angles used to analyze posture throughout the stride.

Statistical Analysis. The large statistical file containing the key variables describing the gait patterns of all the volunteers was transferred to a VAX 780 main-frame computer where programs from BMDP (Berkeley, CA) were used for statistical comparisons between the different experimental conditions. Means and standard deviations for each variable under each testing condition were calculated. A 2-way analysis of variance with repeated measures was performed on each of the variables using the BMDP 2V program, with 3 levels of speed and 4 levels of load. Post-Hoc Tukey tests were employed to locate the differences between treatment means when significant treatment effects were found by analysis of variance.

Results and Discussion

Test Volunteer Characteristics

The test volunteers were all physically fit males, a bit above average in both height and body mass (Table 1). All engaged in regular physical activity. Of the 16 volunteers, 11 were enlisted U.S. Army personnel, three were Army officers, and two were civilian employees of the U.S. Army Research Institute of Environmental Medicine.

Table 1. Physical characteristics of the test volunteers (means +/- SD)

Age (yr)	30.3 +/- 9.2
Height (cm)	181.2 +/- 7.5
Body mass (kg)	76.8 +/-8.9
Gender	all male
n	16

Load Effects

We found several statistically significant (p<0.05) effects of backpack load on gait biomechanics. The following is a summary of the major load effects, descriptions of which can be found in much greater detail in our technical report on the effects of backpack weight on gait biomechanics (19).

The following increased significantly with increasing load:

> stride frequency
> time of toe-off as % of stride
> percentage of stride under double-support
> minimum knee angle
> hip range of motion
> forward trunk inclination
> trunk range of motion
> minimum horizontal velocity
> propulsive impulse
> peak and average propulsive force
> % of stride at peak propulsive force
> braking impulse
> peak and average braking force
> lateral impulse
> average lateral force
> medial impulse
> peak and average medial force
> vertical impulse
> average vertical force
> most of the peak bone-on-bone forces at the ankle, knee, and hip

most of the peak muscle torques about the ankle, knee, and hip
electrical activity of the trapezius, quadriceps, hamstrings, tibialis anterior, and gastrocnemius.
It is noteworthy that the electrical activity of the spinal erectors decreased when the load increased from 6 to 20 kg, and only exceeded electrical activity at the 6 kg load when the load increased to 47 kg. This is likely related to the postural adjustments made with the different loads.
peak downward and backward backpack acceleration

The following decreased significantly as the backpack load increased:

stride time
knee range of motion
minimum hip angle
maximum hip angle
degree of rearward arm swing
degree of forward arm swing
shoulder swing range of motion
maximum vertical position
minimum vertical position

Within the range of walking speeds tested, the adjustments to increasing backpack load were consistent. Stride frequency increased as stride length tended to drop. Each foot stayed on the ground for a greater percentage of the stride, through increased hip range of motion, thereby increasing the percentage of the stride in double support. Arm swing decreased in both the forward and backward directions. The body as a whole stayed lower, mainly due to increased forward inclination of the trunk. With increasing backpack weight, the body didn't slow down as much when the foot contacted the ground. These changes in gait with increased load can for the most part be regarded as positive adaptations. However, the increase in forces and torques at the ankle, knee, and hip, an inescapable consequence of carrying heavy loads, most likely increases the risk of musculoskeletal injury.

Speed Effects

There were three trials at each combination of load and speed, and preliminary film analysis was used to select the trial at each condition that came closest to the nominal speed. The volunteers deviated somewhat from the visually cued walking speeds of 1.1, 1.3 and 1.5 meters per second. They apparently had difficulty keeping their walking speed down to the slowest experimental pace of 1.1 m/s. Of the set of trials selected for final analysis, it was found that the volunteers cued to walk at the slowest speed of 1.1 m/s actually walked at 1.17 ± 0.06 m/s. The volunteers cued to walk at the medium speed of 1.3 m/s walked only slightly faster than the cued pace, at 1.33 ± 0.05 m/s. The volunteers cued to walk at the fast speed of 1.5 m/s were right on target, actually walking at 1.50 ± 0.06 m/s. Because the actual walking speeds deviated from the cued speeds, the means of the three walking speeds differed by about 0.17 m/s instead of the planned 0.20 m/s,. Thus, the increases in speed from slow to medium to fast were in steps of about 13%-14% instead of the planned 15%-18%. Even though the walking speeds were not exactly as intended, they still corresponded to slow, medium, and fast load carriage speeds and likely represented a natural range of speeds for soldiers marching with backpack loads. Subsequent to this experiment we added an electric-eye speed trap to the system so that trials that deviated by more than 5% from the target speed could be rejected.

We found several statistically significant ($p < 0.05$) effects of walking speed on gait biomechanics. The following is a summary of the major load effects, descriptions of which can be found in much greater detail in our technical report on the effects of walking speed on the biomechanics of load carriage (20).

The following increased significantly as load carriage walking speed increased:

> stride length
> stride frequency
> maximum ankle angle
> ankle range of motion
> maximum hip angle
> hip range of motion
> maximum shoulder angle
> shoulder swing range of motion
> maximum upward velocity of the body center of mass
> maximum downward velocity of the body center of mass
> vertical range of motion of the body center of mass
> propulsive impulse
> braking impulse
> lateral impulse
> average propulsive force
> average braking force
> peak propulsive force
> peak braking force
> peak lateral force
> peak upward-downward and forward-backward bone-on-bone ankle forces
> peak upward-downward and forward-backward bone-on-bone knee forces
> peak upward-downward and forward-backward bone-on-bone hip forces
> peak ankle dorsiflexion torque
> peak knee extension torque
> peak hip flexion and extension torque
> electrical activity of the trapezius, spinal erectors, quadriceps, hamstrings, tibialis anterior, and gastrocnemius
> peak upward, downward, and backward backpack acceleration

The following decreased significantly as load carriage walking speed increased:

> stride time
> time of toe-off as % of stride
> percentage of stride under double-support
> minimum hip angle
> minimum elbow angle
> degree of rearward arm swing
> minimum vertical position of the body center of mass
> medial impulse
> vertical impulse
> time of peak propulsive force as % of stride

It is noteworthy that trunk range of motion did not change at all with increases in walking speed. Also, average vertical force exerted by the foot on the ground increased less than 1% as walking speed increased 28% from the slowest to the fastest pace.

The adjustments to increased walking speed were consistent over the range of the backpack loads tested.
The 14% jumps in speed from 1.17 to 1.33 m/s and from 1.33 to 1.50 m/s were accompanied by 6-7% jumps in both stride length and stride frequency. The longer stride was effected both by reaching out further forward with the leg and pushing further backward with it, necessitating greater hip and ankle range of motion. This was accompanied by very large increases in hip extension and knee extension torque as well as large increases in hip flexion torque. Peak propulsive force occurred at an earlier percentage of stride. The importance of muscular work in extending the hips and knees to increasing walking speed was evidenced by an 83% increase in

hamstring electrical activity when going from the slowest to the fastest walking speed and a 40% increase in quadriceps electrical activity. All of the other muscles monitored increased in their electrical activity as well, although to a lesser degree. As the legs stretched apart during the longer stride, the body's center of mass dropped lower, thus traveling through a greater vertical excursion. Upward and downward velocity of the body increased. The degree of arm swing increased both towards the front and the back of the body, and the elbow bent more. It is important to note that because the toe lifted off the ground at an earlier percent of stride, the percentage of stride in double-support decreased, an effect opposite to that brought about by increasing the load. Increases in walking speed were brought about more by increases in horizontal than vertical forces. While propulsive, braking, and lateral impulses increased with walking speed, vertical impulse actually decreased. Average propulsive and braking forces increased over 20% from the slowest to the fastest walking speeds, but vertical force increased less than 1%. Despite the lack of increase in vertical ground reaction force with increasing walking speed, bone-on-bone forces increased in both the vertical and horizontal directions. With the backpack tested, peak accelerations of the pack increased with walking speed in all but the forward direction because flexibility in the strap system damped acceleration in that direction.

Combined Effects of Load and Speed

The fact that there were few statistical interaction effects of load and speed means that, for the most part, increases in load had the same effects on gait over the full range of walking speeds tested and increases in speed had the same effects on gait over the full range of backpack loads tested. As a result, the effects of speed and load were relatively uncomplicated. Many of the effects were in the same direction. For example increases in both speed and load resulted in increased joint torques. However, some of the effects of increasing load were opposite in direction to those of increasing speed, so that for certain variables, the effects of speed and load tended to cancel each other out. The following shows which effects were in the same direction for increases in speed and load, and which effects were opposite in direction. These combination effects are sub-categorized into those that have no apparent risk and those with possible attendant risks.

The following increased when speed and load increased, with no obvious attendant risks:

Stride frequency
Hip range of motion

The following increased when speed and load increased, with possible attendant risks:

Bone-on-bone forces and muscle torques: Greater forces pushing the bones together and pulling them apart probably increase the likelihood of injury to bones, articular surfaces, and ligaments. The greater muscle torques can be expected to increase the likelihood of muscle and whole-body fatigue and injury to muscles and tendons.
Propulsive, braking, and lateral impulses: As the product of force and time, impulse may be associated with muscle fatigue and injury to various tissues.
Propulsive, braking, and lateral forces: While we can't always determine whether an injury is the result of a single large force or repeated applications of smaller forces, higher force is more likely to result in tissue injury.
Downward and backward backpack acceleration: Acceleration is the result of force. Force on the backpack can be attributed to either gravity or the force exerted by the torso on the pack. Greater acceleration of the pack suggests greater reaction forces of the pack on the torso, applied to the shoulder straps or hip pad and belt, which may increase the likelihood of discomfort or injury to skin, nerves, and blood vessels.
Muscle electrical activity: Increases in muscle electrical activity are associated with greater force generation, which are associated in turn with increased fatigue and injury risk.

The following decreased when speed and load increased, with no obvious attendant risks:

> Minimum vertical position
> Minimum hip angle

There were no variables which both decreased when either speed or load increased and resulted in apparent attendant risk.

The following change in opposite directions with increases in speed and load, with no obvious attendant risks:

> Vertical impulse: Impulse, as the product of force and time, increases if either load or time increases. Higher backpack loads increase vertical impulse by increasing force, while increased walking speed reduces vertical impulse by shortening stride time.

> Arm swing: With increased load, the degree of arm swing lessens. Arm swing helps keep the torso from rotating excessively during walking by applying the increase in body angular momentum in the transverse plane, caused by off-center foot push-off forces, to the arms rather than to the torso. When the pack becomes heavier, it increases the inertia of the pack-torso combination. Angular momentum is the product of speed and inertia. Thus, since pack-torso inertia increases, the velocity of the torso for a given angular momentum decreases, reducing the need for arm swing to limit rotation of the torso. Increases in walking speed are effected by greater propulsive forces by the feet, which impart greater angular momentum to the body, in turn increasing arm swing for the reasons cited above. Thus, increased load and increased speed have opposite effects on arm swing. However, this has no apparent negative consequences.

The following changed in opposite directions when speed and load increased, with possible attendant risks:

> Percentage of stride in double-support and time of toe-off as percent of stride: An increase in these variables is considered a positive adaptation to increased load because it provides more stability and may reduce stress on the musculoskeletal system. However, as walking speed increases these measures decrease, tending to cancel the potential positive adaptations to increased load.

> Percent of stride at peak propulsive force: This measure increases as the load increases and decreases as the speed increases. The later occurrence of peak force as load increases may relate to earlier placement of the foot on the ground to increase double support time. The decrease in this measure with increased walking speed may represent a negation of this positive adaptation.

> Maximum hip angle: A decrease in this measure is related to a shortened stride and quicker cadence at increased load, a positive adaptation because it provides more stability and may reduce stress on the musculoskeletal system. However, increased walking speed counteracts this effect by lengthening the stride and increasing the hip angle as the foot pushes off, with possible increased risk.

Statistical Interactions of Speed and Load

There were a few variables exhibiting statistical interaction. That means that the effects of increasing speed were not the same for all loads and the effects of increasing load were not the same for all speeds. The variables showing such statistical interaction were:

Stride frequency: At the 1.17 and 1.33 m/s walking speeds, stride frequency increased markedly when the load increased from 33 to 47 kg. No such adaptation occurred at the 1.50 m/s walking speed.

Stride time: At the 1.17 and 1.33 m/s walking speeds, stride time decreased markedly when the load increased from 33 to 47 kg. No such adaptation occurred at the 1.50 m/s walking speed.

Downward impulses for shank-on-foot, thigh-on-shank, and trunk-on-thigh: Impulse, the area under the force vs. time curve over a full stride, increased with load carried, but decreased with increasing speed as stride time became shorter. The statistical interaction was due to the fact that the increase in impulse was directly related to pack weight except for the 1.33 m/s walking speed, for which the increase in impulse was less than proportional to the increase in pack weight when going from the 33 kg to the 47 kg pack.

All the above variables are related, accounting for the fact that they all showed statistical interaction in their responses to load and speed. Stride time and stride frequency are the mathematical inverses of each other. Since impulse is the product of force and time, impulses at the ankle, knee, and hip are sensitive to changes in stride time. The statistical interactions of stride frequency and stride time are related to the fact that, at the 1.17 and 1.33 m/s speeds, volunteers adapted to the heaviest load by taking shorter steps at a more rapid cadence, thus maintaining a stable base of support and avoiding excessive impulse about the lower limb joints. Yet this did not occur at 1.5 m/s, showing a lack of impulse-reducing gait adaptation to the heaviest load when walking at the fastest speed. The statistical interaction of the impulse variables is related to the fact that, at the 1.33 m/s walking speed, a gait adaptation occurred that didn't increase impulse as much as expected when increasing from the 33 kg to the 47 kg backpack. This adaptation did not occur at the 1.50 m/s walking speed. The lack of adaptation is likely related to the fact that, at the fastest walking speed, the volunteers could not further increase their stride frequency in order to shorten stride time. Without a reduction in stride time, an impulse increase could not be moderated.

Conclusions

It is clear from the study results that increasing either load or speed results in increased stress to the musculoskeletal system, as evidenced by bone-on-bone forces, ground reaction forces and impulses, and muscle electrical activity, which most probably increases the rate of fatigue and risk of injury. These effects are for the most part additive, as evidenced by the small percentage of variables showing statistical interaction, indicating that the effects of increased load are the same regardless of walking speed, and the effects of increased walking speed are the same regardless of backpack load. Thus, the combination of fast walking and heavy load can present a relatively high level of risk for fatigue and injury. The few variables that showed statistical interaction provided even more evidence that the combination of fast walking speed and heavy load can be particularly risky. At the fastest walking speed, 1.5 m/s (5.4 km/hr, 3.4 mi/hr), the volunteers could not shorten their stride length and increase their stride frequency when the load increased from 33 kg to 47 kg as they did at the slow and medium walking speeds (1.17 and 1.33 m/s). Their inability at the fast walking speed to make this adaptation to increased load means they could not effectively increase their stability and reduce the potential stresses to their legs and feet. It thus appears prudent to recommend that soldiers should avoid, if possible, walking faster than 1.33 m/s (4.8 km/hr; 3.0 mi/hr) when carrying backpack loads in the vicinity of 45 kg (100 lb).

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